Closed-Loop Control of Functional Neuromuscular Stimulation

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1. SYNTHESIS OF UPPER EXTREMITY FUNCTION

The overall goals of this project are to (1) measure the biomechanical properties of the neuroprosthesis user's upper extremity and incorporate those measurements into a complete model with robust predictive capability, and (2) use the predictions of the model to improve the grasp output of the hand neuroprosthesis for individual users.

1. a. BIOMECHANICAL MODELING: PARAMETERIZATION AND VALIDATION

Purpose

In this section of the contract, we will develop methods for obtaining biomechanical data from individual persons. Individualized data will form the basis for model-assisted implementation of upper extremity FNS. Using individualized biomechanical models, specific treatment procedures will be evaluated for individuals. The person-specific parameters of interest are tendon moment arms and lines of action, passive moments, and maximum active joint moments. Passive moments will be decomposed into components arising from stiffness inherent to a joint and from passive stretching of muscle-tendon units that cross one or more joints.

Progress Report

1. a. i. MOMENT ARMS VIA MAGNETIC RESONANCE IMAGING

Abstract

No activity took place with regard to this project this quarter.

1.a.ii. PASSIVE AND ACTIVE MOMENTS

Abstract

In order to verify that our analytical model of the passive moments about the finger is accurately portraying the mechanics at the finger joint, we deemed it necessary to produce a mechanical model of the finger with which to test our data collection and analysis .

Purpose

The purpose of this project is to characterize the passive properties of normal and paralyzed hands. This information will be used to determine methods of improving hand grasp and hand posture in FES systems.

Progress Report

During this quarter, we developed a mechanical model of the passive properties of the finger joint to verify that the mathematical model behaved in the manner we expected. We developed the model to most closely match the data which we had collected from the MP joint of the index finger, while the IP joints of the finger were constrained. The model also needed to fit into the constraints of the measurement apparatus.

The model we developed was based on two springs as shown in Figure 1.a.ii.1. The two springs are connected to a pulley by a swivel joint. When a torque is applied to the rod, resistance is applied by the springs. Each spring represents both flexion and extension, depending upon which side of neutral the rod is rotated. The balance between flexion and extension resistance can be adjusted by the lateral location of the spring. The overall tension of each spring can be adjusted by the distance between the spring and the pulley. The extrinsic spring can be adjusted to a number of repeatable locations to represent the change in moment arm as the wrist is flexed or extended.

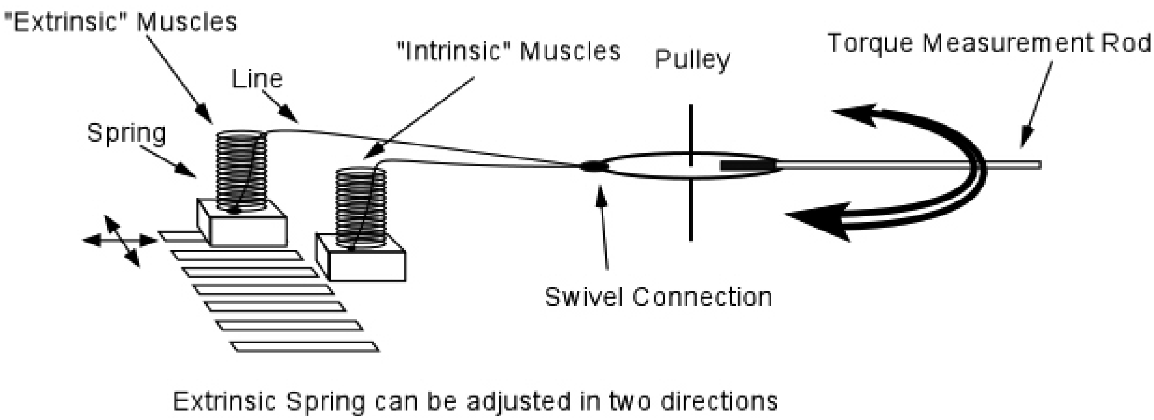


Figure 1.a.ii. 1. Two spring model of the passive moments at the MP joint of the finger. The springs are allowed to bend, but not extend.

The output of the torque measurement apparatus is shown in Figure 1.a.ii.2. The torques for intrinsic and extrinsic forces are shown separately and summed together. In Figure 1.a.ii.3. measurements are displayed for three separate wrist angles, showing the effect of the extrinsic forces on the passive moments.

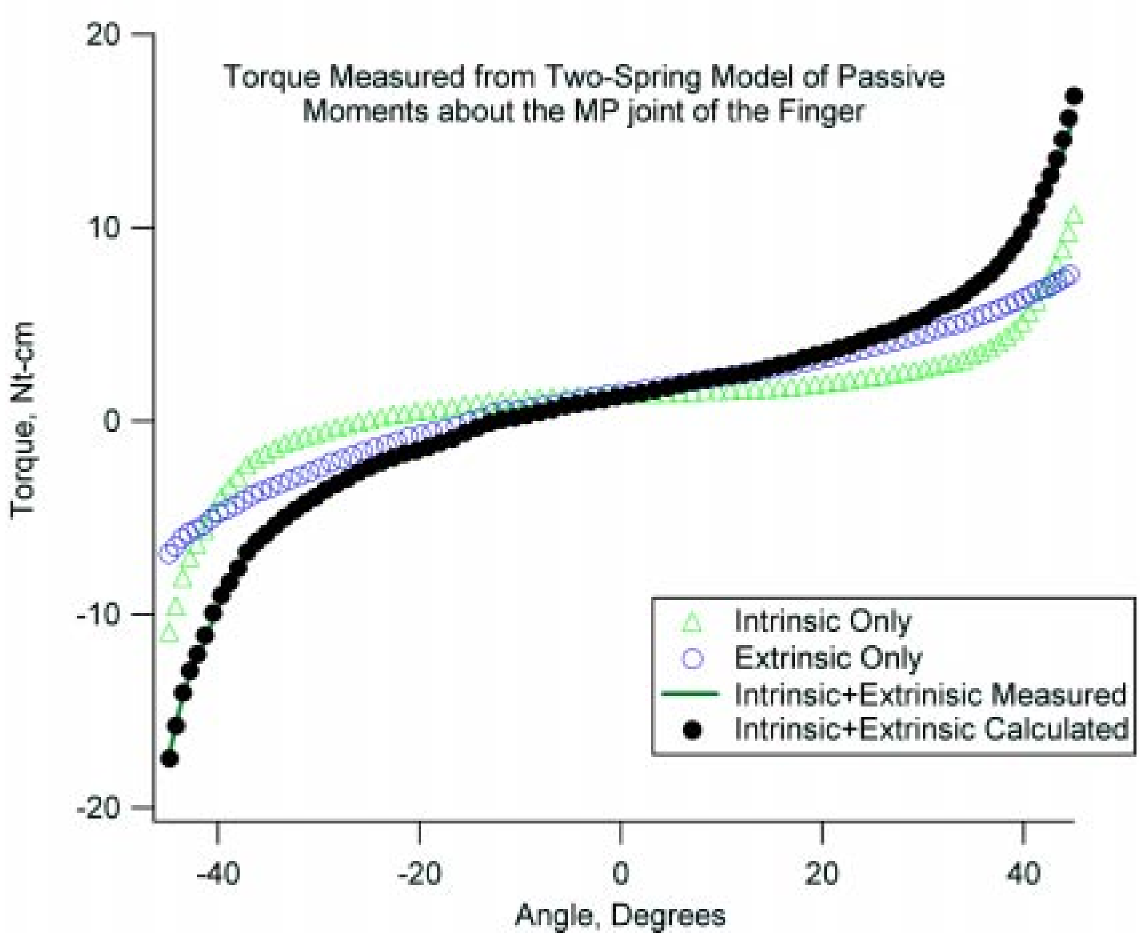


Figure 1.a.ii. 2. Moment Angle Curves of the Finger model using springs for the intrinsic and extrinsic resistances. The Calculated torques are a sum of the measured intrinsic and extrinsic forces.

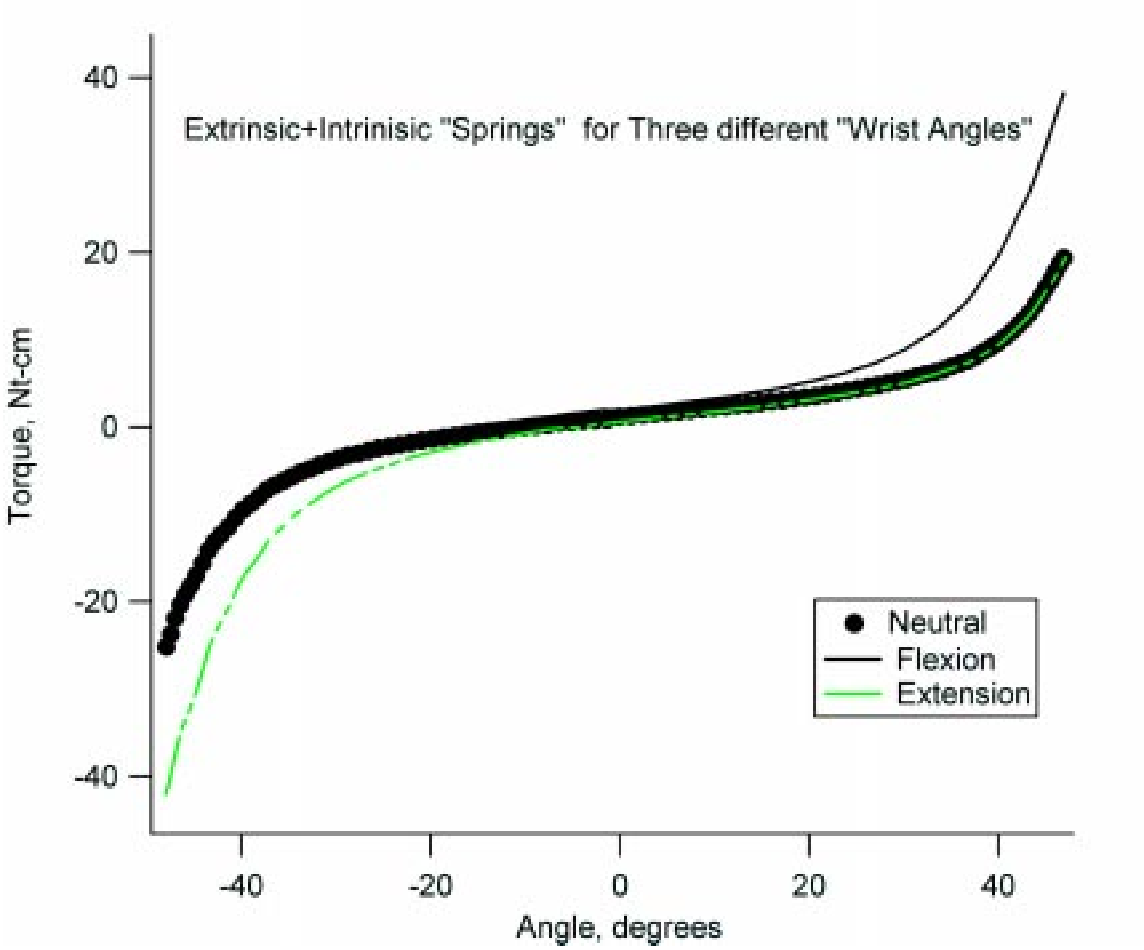


Figure 1.a.ii.3. Moment Angle Curves of the Finger model using springs for the intrinsic and extrinsic resistances. The curves for three wrist angles are shown.

Plans for Next Quarter

Next quarter we plan to use the mechanical model to verify the analytical model and make appropriate adjustments to the analytical model as needed.

1. b. BIOMECHANICAL MODELING: ANALYSIS AND IMPROVEMENT OF GRASP OUTPUT

Abstract

Our biomechanical model of the Br-ECRB tendon transfer and preliminary clinical assessments of wrist function indicate that the ability to voluntarily extend the wrist depends on the position of the elbow after the tendon transfer. To evaluate the influence of biomechanics in determining function, we measured the active range of motion at the wrist, passive range of motion at the wrist, lateral pinch strength, voluntary wrist extension strength, and voluntary elbow extension strength in six limbs (from five individuals) with Br-ECRB tendon transfers. This progress report summarizes the active range of motion and voluntary wrist extension strength measured in all six subjects.

Purpose

The purpose of this project is to use the biomechanical model and the parameters measured for individual neuroprosthesis users to analyze and refine their neuroprosthetic grasp patterns.

In the past quarter, we have evaluated how the passive moment-generating capacity of the tight and slack Br-ECRB transfer (described in previous progress reports) influences gravity-assisted wrist flexion. The net passive moment at the wrist joint (before a Br-ECRB transfer) was compared to the passive wrist extension moment generated by the transfer to estimate the range of wrist postures where gravity-assisted wrist flexion is possible.

Progress Report

In the past quarter, we completed clinical assessments of wrist function and quantitative measurements of voluntary wrist extension strength and elbow extension strengh in five limbs (from four individuals) with Br-ECRB transfers. We have now studied a total of 6 wrists with Br-ECRB transfers, including the data from the subject described in the last QPR.

Wrist range of motion after Br-ECRB tendon transfer

In four of six wrists evaluated, we observed that the maximum wrist position that could be actively maintained against gravity was more extended when the elbow was extended compared to when the elbow was flexed (Fig. 1.b.1). On average, the maximum position was 22.5˚ wrist extension (range = 0˚-42˚) when the elbow was fully extended, and -1.2˚ wrist extension (range = -46˚- -34˚) when the elbow was flexed (120˚ elbow flexion). The rest position of the wrist against gravity also depended on elbow posture. In each of the six wrists evaluated, the rest position of the wrist was more flexed when the elbow was flexed compared to when it was extended. On average, the rest position of the wrist was 28.5˚ wrist flexion when the elbow was extended and was 47.2˚ wrist flexion when the elbow was flexed.

Wrist strength after Br-ECRB tendon transfer

We measured the isometric wrist extension moment generated during maximal effort using the wrist moment transducer as described in the previous QPR. The data will be used to test the hypothesis that wrist extension strength is weaker when the elbow is flexed after the Br-ECRB tendon transfer. We measured isometric wrist extension moment in two elbow postures (~20˚ elbow flexion and 120˚ elbow flexion) and three wrist positions (30˚ wrist extension, 0˚ wrist extension (neutral), and 30˚ wrist flexion). For each combination of wrist/elbow position, we collected 4 trials of data. The average of these four trials is reported.

Based on a preliminary analysis of the data, the relationship between wrist extension strength and elbow position depends on the position of the wrist after the Br-ECRB tendon transfer. In four of six wrists evaluated, the isometric wrist extension moment generated during maximal effort was weaker when the elbow was flexed compared to when it was extended in at least one wrist posture (Fig. 1.b.2). In one wrist, wrist extension strength was weaker when the elbow was flexed in all three wrist positions. In two wrists, wrist extension strength with the elbow flexed was either comparable or greater then wrist extension strength when the elbow was extended for all three wrist postures.

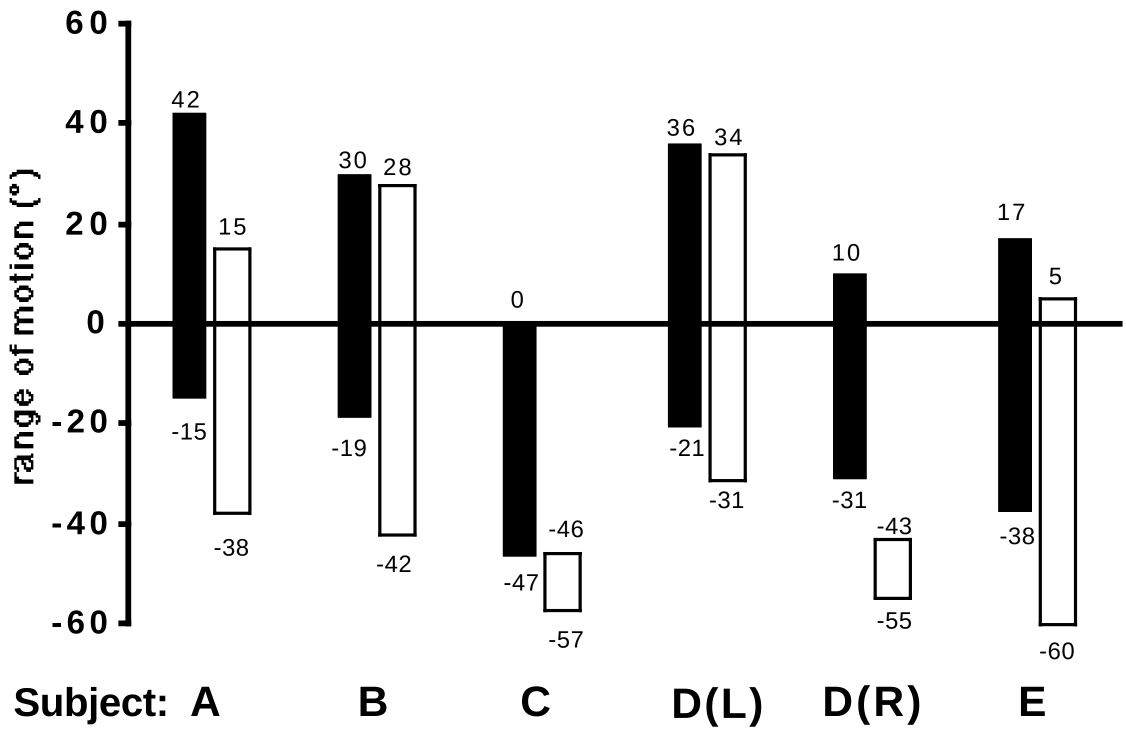


Figure 1.b.1. Active range of motion at the wrist measured in six wrists with Br-ECRB tendon transfers. The top of each bar indicates the maximum position of wrist extension that could be actively maintained against gravity. The bottom of each bar indicates the rest position of the wrist against gravity. Positive numbers indicate wrist extension, negative numbers indicate wrist flexion. Filled bars indicate range of motion measured when the elbow was fully extended (0˚ elbow flexion), open bars indicate range of motion when the elbow was flexed (120˚ elbow flexion). In general, the maximum position of wrist extension was more extended when the elbow was extended. Also, the rest position of the wrist was more flexed when the elbow was flexed.

Summary

Data describing wrist range of motion and voluntary moment-generating capacity has been collected from six wrists (from five individuals) with Br-ECRB tendon transfers. We observed that both active range of motion and wrist extension strength depends on elbow posture in these individuals. This data will be useful in understanding the role of biomechanics in determining wrist function after tendon transfer.

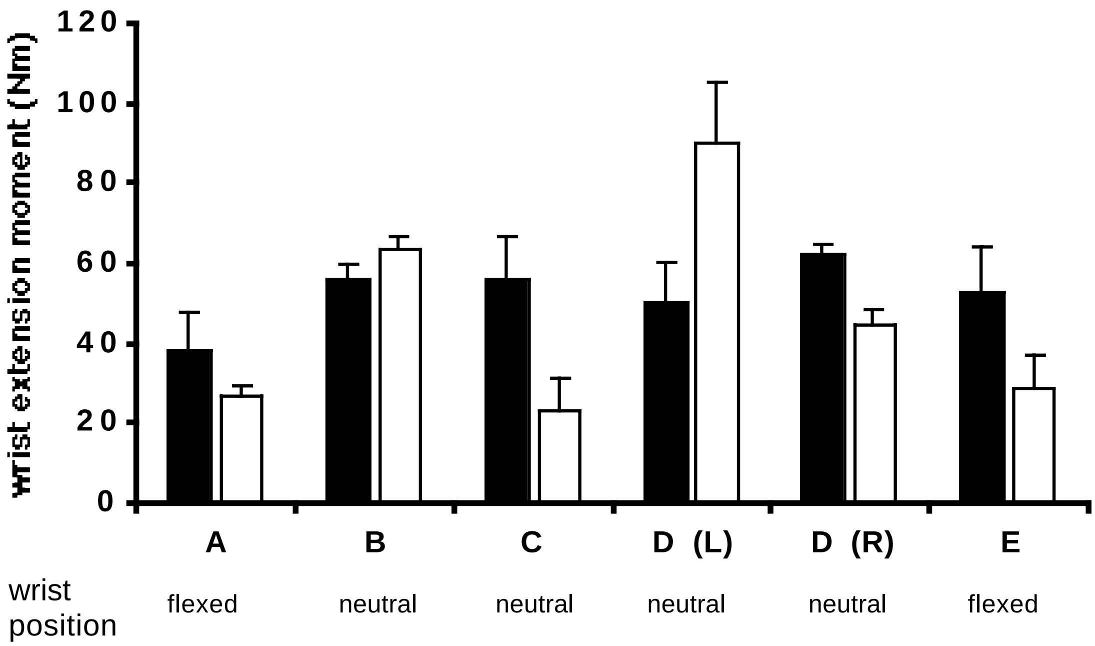


Figure 1.b.2. Isometric wrist extension moment measured when the elbow was extended (filled bars) and when the elbow was flexed (open bars). In four out of the six wrists studied, we observed that wrist extension strength decreased when the elbow was flexed in at least one wrist posture. The greatest measured difference between wrist extension strength in the two elbow postures is shown, and the wrist position is indicated below the graph.

Plans for Next Quarter

In the next quarter we plan to complete a statistical analysis of the data described in this progress report. Measurements of elbow extension strength and the elbow flexion moment generated during wrist extension have been collected in these subjects as well. Analysis of the data collected at the elbow joint will be completed in the next quarter.

2. CONTROL OF UPPER EXTREMITY FUNCTION

Our goal in the five projects in this section is to either assess the utility of or test the feasibility of enhancements to the control strategies and algorithms used presently in the CWRU hand neuroprosthesis. Specifically, we will: (1) determine whether a portable system providing sensory feedback and closed-loop control, albeit with awkward sensors, is viable and beneficial outside of the laboratory, (2) determine whether sensory feedback of grasp force or finger span benefits performance in the presence of natural visual cues, (of particular interest will be the ability of subjects to control their grasp output in the presence of trial-to-trial variations normally associated with grasping objects, and in the presence of longer-term variations such as fatigue), (3) demonstrate the viability and utility of improved command-control algorithms designed to take advantage of forthcoming availability of afferent, cortical or electromyographic signals, (4) demonstrate the feasibility of bimanual neuroprostheses, and (5) integrate the control of wrist position with hand grasp.

2. a. HOME EVALUATION OF CLOSED-LOOP CONTROL AND SENSORY FEEDBACK

Abstract

In this quarter we completed the review and design for the third-generation single-channel grasp force feedback system. The design modifications improved all sections of the device: power, output, pulse generation, and force-to-current conversion. The revised circuitry has been bread-boarded and partially qualified. The circuitry will be constructed into a complete and portable unit in the next quarter.

Purpose

The purpose of this project is to deploy a portable hand grasp neuroprosthesis capable of providing closed-loop control and sensory feedback outside of the laboratory. Our goal is to evaluate whether the additional functions provided by this system benefit hand grasp outside of the laboratory.

Progress Report

The third-generation unit will be constructed in a modular form similar to the original prototype. The single circuit board with hand-soldered surface mount components used in the second prototype conserves space, but repairs are difficult and components cannot be revised practically. The modules in the third unit will comprise (a) the low- and high-voltage power supplies, (b) the pulse generator and output stage, (c) the arithmetic unit, and (d) the interface/connectors. The modular design will allow parts of the circuit to be tested and modified without affecting those parts that work well. For example, modules (a) and (b) have worked well in both previous versions and will be largely unchanged, whereas module (c) will require significant revision, and more than one prototype may be constructed. The overall design and the specific design changes are described below.

POWER

The ±5V supplies worked reasonably well, but the switched inductor DC-DC converters generated switching spikes synchronized to the pulse output. To minimize the switching noise, we replaced the +5V converter with a low power, low dropout, linear regulator that was able to maintain a consistent +5V output for input (battery) voltages as low as +5.6V under the maximum load drawn by the complete stimulator. The regulator also reduced part count and board space which is at a premium in the complete, portable unit. The –5V supply still required a switched converter to invert the one-sided supply, but a switched-capacitor circuit (charge pump) was substituted for the switched inductor circuit used previously to reduce switching noise. The 120V supply was replaced in the revised prototype with a similar, but more robust design, and the design will be carried forward in the new prototype.

OUTPUT STAGE

The output stage works well. The current-regulating diode used in the prior versions (1N5291, i_{reg}=0.56 mA) was replaced, however, with a lower current version (0.22 mA, 1N5283) in order to reduce the output current error at low currents (the output current is equal to the sum of the programmed current and the regulation current). The change was feasible since the pulse period exceeds the time required to recharge the storage capacitor following a pulse. Assuming a worst-case pulse width of 500 µsec at 20 mA, the charge drained from the capacitor will be 10 µC. That charge can be restored in only 45 msec at 0.22 mA — just within the pulse period of 50 msec (20 Hz). The op-amp and transistor were changed slightly to reduce power consumption and improve stability. Last, an LED was added in series with the load to indicate that pulses are being generated. The LED will consume a small amount of additional power, but will be very useful for the user or an attendant to confirm that the device is running. The current through the LED is equal to the output current so its brightness roughly indicates stimulus strength.

PULSE GENERATOR

The pulse generator is based on a dual multistable monovibrator (CD4538). The only change was to connect the pulsewidth resistor via a socket so it can be revised, as needed. The pulsewidth setting is not intended to be user-adjustable, but does need to be changeable by a support engineer in the event stimuli at a conservatively short pulse width are not intense enough. The pulse-rate resistor will also be replaceable, although it is unlikely that the rate will need to be changed.

Since the output stage and pulse generator are relatively mature designs, they will be built into a single module.

ARITHMETIC UNIT

The arithmetic unit (AU) configuration has been revised significantly based on recent calibrations of the FSR response, a correction to the original derivation of the transfer function, and the relatively poor performance realized in the previous units. For the purposes of this discussion, control of the threshold pulse amplitude is considered part of the AU function, rather than the pulse generator function as in previous descriptions.

The most significant revision results from the deviation of the resistance characteristic of the FSR from the nominal power function used to describe it previously. The actual resistance decreased more rapidly with force than desired, yielding strongly decelerating current-versus-force characteristics. The resistance relationship is improved by placing a resistor R_p in parallel with the FSR, but at the expense of a compression of the total range of resistance. Mathematically, the resistance characteristic (with R_p in place) is fit better by the power function:

$$R_{FSR} \| R_P = R_{\min} \left(F / F_{\min} \right)^b \qquad (1)$$

where R_{min} is the resistance at a defined minimum force F_{min}, typically 2N. However, the exponent b is relatively small, roughly 0.2 — 0.3. As a consequence, the arithmetic unit had to be revised so that the net current-versus-force characteristic has a power function exponent m in the range of 0.35 — 0.65. That exponent is the product of the FSR exponent b and the programmed exponent of the AU, a., which must range roughly from 1 to 3. The AU was configured previously to produce exponents less than 1. The change requires a different pin connection on the AU. The exponent is also determined by the resistance of that connection, R_A, according to the relation:

$$a = \frac{196\ \Omega}{R_A} + 1 \qquad (2)$$

The inverse relationship poses a difficulty since linear changes in RA produce highly nonlinear changes in the exponent. The original design using a linear taper potentiometer for user adjustment of the exponent is infeasible since equal increments in the potentiometer position produced large very large changes in exponent over a range of small R_A, and negligible changes for modest to large values of R_A. We compensated for the nonlinearity by using a log taper potentiometer (500 Ω) in series with a small resistor (90 Ω) to both limit and linearize the changes in the exponent with potentiometer position.

The last change to the AU was to an exponential amplifier to adjust the amplitude of the stimulus pulse produced at the minimum criterion force. The purpose of the amplifier is to transform linear changes in the adjustment potentiometer position to ratiometric changes in the pulse amplitude. The same potentiometer will be equipped with the power switch, so the device will always be turned on at zero stimulus amplitude.

Overall, the circuit revisions will improve the user adjustment of the exponent and minimum stimulus amplitude greatly by making the perceptual changes in the output commensurate with or proportional to changes in the potentiometer positions. The output will also be more "proportional" (logarithmically) to force throughout the force range.

Plans for Next Quarter

The design changes will be executed as the third prototype is constructed. The anticipated device characteristics will be validated through bench testing and limited user testing.

2. b. INNOVATIVE METHODS OF CONTROL AND SENSORY FEEDBACK

2. b. i. ASSESSMENT OF SENSORY FEEDBACK IN THE PRESENCE OF VISION

Abstract

This project is now completed. The user manual for the video simulation system has been completed (see 2.b.ii).

Purpose

The purpose of this project is to develop a method for including realistic visual information while presenting other feedback information simultaneously, and to assess the impact of feedback on grasp performance. Vision may supply enough sensory information to obviate the need for supplemental proprioceptive information via electrocutaneous stimulation. Therefore, it is essential to quantify the relative contributions of both sources of information.

2. b. ii. INNOVATIVE METHODS OF COMMAND CONTROL

Abstract

The purpose of this project is to develop new command control algorithms that will make control of neural prosthetic hand grasp simpler and more effective. During this quarter, continued analysis was carried out with all of the data taken while testing the command control algorithms. The command control algorithm study and results were submitted in abstract form for the annual RESNA conference and a complete publication is in preparation. Finally, documentation of the video-based simulator, command control testing was completed and a user's manual was written to facilitate future use of the computer-based video simulator.

Purpose

The purpose of this project is to improve the function of the upper extremity hand grasp neuroprosthesis by improving user command control. We are specifically interested in designing algorithms that can take advantage of promising developments in (and forthcoming availability of) alternative command signal sources such as EMG, and afferent and cortical recordings. The specific objectives are to identify and evaluate alternative sources of logical command control signals, to develop new hand grasp command control algorithms, to evaluate the performance of new command control sources and algorithms with a computer-based video simulator, and to evaluate neuroprosthesis user performance with the most promising hand grasp controllers and command control sources.

Progress Report

1. Data Analysis

Further analysis has been conducted on the data gathered from the command control algorithm testing. This analysis sought to find a pattern of success rates among the variables. This analysis has been focused on the results (high, low, or success) from the individual phases (acquire, hold, and modify) of each trial. The variables considered when analyzing the data included 5 subjects, 3 window sizes, 7 algorithms, 2 data sets, 3 to 4 sessions in each data set, 15 trials per block of trials, and 3 phases per trial. No patterns of statistical or physical significance have yet been identified, and this analysis will continue in the coming quarter.

2. Conference/Abstract-Publication

The data collected during the testing of the new command control algorithms, along with the subsequent statistical analysis, were submitted in abstract form to the Rehabilitation Engineering Society of North America (RESNA) 2000 Annual Conference, where they will be presented in July.. The results are also being prepared for submission to a peer-reviewed archival journal.

3. User's Guide

A user's guide to the video simulator and evaluation system has been written. This guide provides basic instructions for system setup and startup, running trials, and saving data. The manual also contains a documented copy of the programming code for the simulator and evaluation software. The code allows advanced readers to gain a better understanding for how the simulator works in case modifications are desired.

Plans for Next Quarter

In the next quarter, we expect to complete the phase-based analysis of the data from command control evaluation experiments and complete preparation of a manuscript reporting our results.

2. b. iii . INCREASING WORKSPACE AND REPERTOIRE WITH BIMANUAL HAND GRASP

Abstract

Testing and evaluation of the EEG biopotential interface to the hand grasp neuroprosthesis was conducted this quarter. Results indicate that it can be used to control the hand grasp system, however the speed is inferior to the existing controllers. Limitations are discussed and plans for possible future direction are provided.

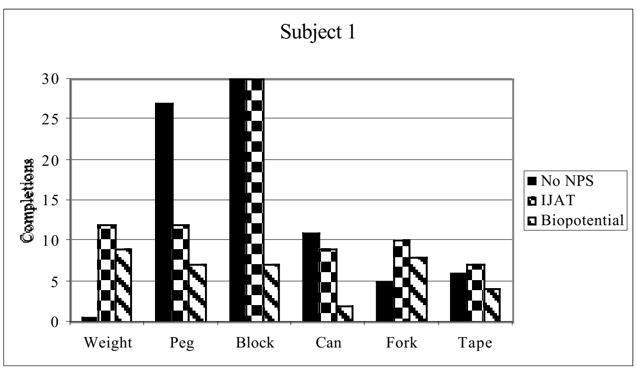
Purpose

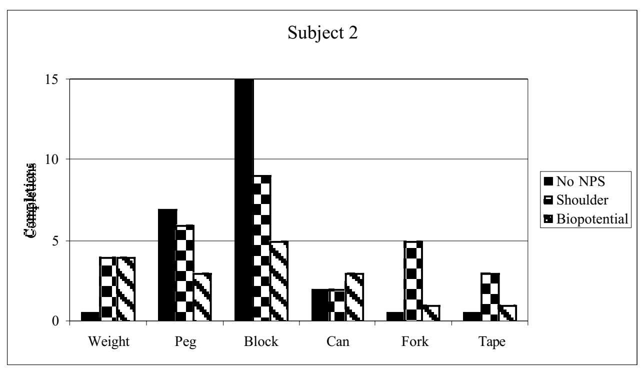
The objective of this study is to extend the functional capabilities of the person who has sustained spinal cord injury and has tetraplegia at the C5 and C6 level by providing the ability to grasp and release with both hands. As an important functional complement, we will also provide improved finger extension in one or both hands by implantation and stimulation of the intrinsic finger muscles. Bimanual grasp is expected to provide these individuals with the ability to perform over a greater working volume, to perform more tasks more efficiently than they can with a single neuroprosthesis, and to perform tasks they cannot do at all unimanually.

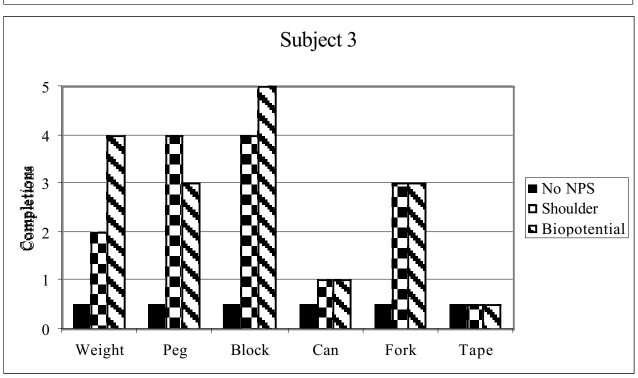
Progress Report

During the last quarter, a final set of analyses into the assessment of the interface to the hand grasp neuroprosthesis was conducted and results of one subject were reported. Two additional neuroprosthesis users were trained on the control of the 25-29 Hz signal recorded from the frontal areas, and when then tested on using the interface with the Grasp and Release Test (GRT). The algorithm used to convert the signal into a command to the neuroprosthesis was the 'hold switch' algorithm, discussed in the previous report. As a review, the operation of the algorithm is as follows. In the operation of this program, a suppression of the signal below a set threshold initiated a ramp signal to close the hand at a fixed rate until full hand closure was achieved. The command to maintain the hand in a closed posture was only achieved while the signal was below the threshold. When the signal returned to above the threshold value, the command signal was reversed and the hand opened at a fixed rate until full hand opening was achieved. To prevent inadvertent changes to the command signal, a delay of 200 ms, or two sampling periods, was introduced to prevent spontaneous signal spikes from generating a command signal.

The results of the GRT assessment are represented in Figures 2.b.iii.1-2.b.iii.3 for all three subjects. In each of the figures, the mean number of completions across the three trails is shown for each object. Also shown in the graphs are the results on the GRT when the same subject used their existing controller to operate the neuroprosthesis. In this case, Subject #1 was using wrist position as the command control input (derived from the implanted joint angle transducer [IJAT] in the wrist), while Subjects 2 and 3 were using the standard shoulder controller. Also shown in these figures are the results on the GRT for each subject when the neuroprosthesis was not in operation.







Figures 2.b.iii.1 through 2.b.iii.3 – Results of the GRT assessment of the EEG or biopotential interface for all three subject studied.

For Subjects 1 and 2, there was a significant ($p < 0.01$) decrease in the number of completions across objects on the GRT when comparing the biopotential interface to the controller used in daily functions. On average, this decrease was on the order of 50 to 70%. However, it can be seen that across objects, independent of size and weight, the number of completions with the biopotential interface was constant for these two subjects, with six completions on average per trial. In Subject #3, there was no significant change in performance between the use of the shoulder controller and the new interface ($p > 0.4$). These results are misleading, however, since the subject performs rather poorly on the test when compared to the other subjects due to contractures of the finger flexors that prevented full hand opening. In spite of the poor performance, Subject #3 also demonstrated consistency across objects in the number of completions achieved, regardless of size and weight. However, in this case the number of completions was approximately four, compared to the six achieved by the other two subjects.

In addition to the number of completions across objects, the GRT was also used to record the number of failures. A failure in this case indicated dropping the object, failure to acquire the object with the correct grasp or in the environment, or a failure to place and release the object in the designated area. For each of the subjects, the number of failures across objects with the standard controllers was approximately 1 or 2, depending upon the object. For all subjects, however, the number of failures with the new interface on the same objects did not change ($p < 0.01$). Therefore, while the use of the biopotential interface may slow user performance, did not make performance any worse.

As stated in the previous report, there were several causes for the delay in the interface that resulted in the poor performance on the GRT test. These were both due to the signal processing steps involved, as well as a physiologic delay between the time that a command change was desired and the signal responded. Because of these delays, it was decided to perform one additional test in which the everyday operation of the interface was assessed. This was accomplished using an activities of daily living (ADL) analysis. In these tests, one subject was asked to perform one of the following tasks: drinking out of a glass, eating with a fork, writing with a pen, using a desk phone, and using a disk drive. None of the objects were modified, and the subject was graded on the level of assistance needed to complete each phase of the task. Each of the tasks was repeated for a total of three trials, and the time to complete a trial was recorded. Although the grading on the levels of assistance for each aspect of the task will be unchanged since the neuroprosthesis was used in both cases, the time to complete each task can be compared for the new interface and the standard controller which could then be used to assess feasibility.

The results of this assessment are shown in Figure 2.b.iii.4. In the graph, a comparison of the mean time to complete a given task is given for the biopotential interface and for the standard controller for this subject. As stated above, the level of assistance necessary to complete each phase of the task was recorded, but this did not change because the neuroprosthesis was used in all cases. From the figure, it can be seen that the time to complete a task was increased slightly when the biopotential interface was used, however, this increase was not significant ($p > 0.3$). The figure also indicates the possible advantage of an interface that does not depend upon joint position. The time to complete the task involving the use of the disk drive was two times greater for the IJAT than for the new interface. This is because the subject had to maintain the wrist in the extended position to close the hand on the disk to remove it from the drive. However, it was difficult for this subject to maintain that wrist position and pull the disk out at the same time. The biopotential interface, because it does not rely upon existing voluntary movement, was not restricted in this case and the subject could remove the disk with ease.

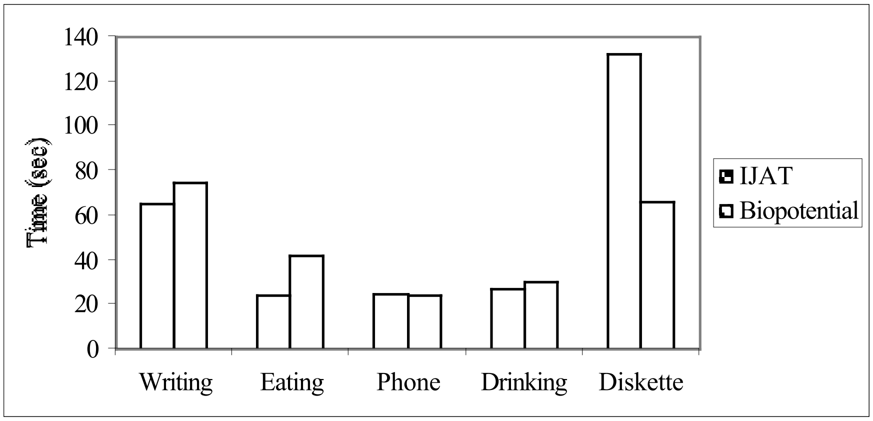


Figure 2.b.iii.4. ADL analysis comparing the biopotential interface to the IJAT system in Subject 1.

The objective of this study was to determine if the EEG signal could be used for the control of a hand grasp system used to restore function to individuals with a cervical level spinal cord injury. It was believed that this alternative method of controlling the system could then be useful for the bilateral implementation of the system. To this end, a collaboration with Dr. Jonathan Wolpaw and colleagues (Wadsworth Center, Albany NY) was instituted to implement these studies. The results from this study indicate that a great deal of work needs to be accomplished if the EEG signal is to be considered for use with the neuroprosthesis. Early results indicated that due to the specific operating parameters of the neuroprosthesis, the EEG signals which were used for the communication devices developed by Wolpaw were not readily applicable. To attempt to address this problem, the EEG signal recorded from the frontal areas was used. However, even after much study, there is still an uncertain amount of contamination in the signal from EMG from the cranio-facial muscles that have made the classification of the signal difficult. Because of this reason, we refer to the signal as a biopotential signal and the interface that was developed as a biopotential interface. However, the principles behind the operation of the biopotential interface are such that any EEG rhythm, such as the mu and central beta studied by Dr. Wolpaw, can be used to operate the hand grasp system if the other problems are addressed.

A better method of recording cortical activity and implementing this for control of the neuroprosthesis will be the use of implanted electrode arrays. In this manner, there is no concern about the nature of the signal. Also, by moving into the cortex, the delays in the signal can be reduced and the information content increased, both of which are additional obstacles in the use of the EEG signal for neuroprosthetic operation. However, the results of this study demonstrate that signals recorded from the cranio-facial areas can be used for the operation of the hand grasp system, independent of the underlying nature of the signal. This information is valuable in that it will allow for the bilateral implementation of the hand grasp system, or for the implementation of a neuroprosthesis in individuals with higher levels of spinal cord injury. It has already been proposed that a new interface can be developed which is capable of recording and differentiating between muscle activity, eye movement, and cortical activity. This will allow for the control of a hand grasp system, using a minimal number (approximately 3) electrodes placed on the frontal areas. Such an interface would then allow for the control of all functions of the neuroprosthesis, without the need for any remaining extremity movement. This frees up the arms for bilateral control, or allows individuals with injuries above the fourth cervical level to benefit from a neuroprosthesis.

Plans for Next Quarter

During the next quarter, the results of these studies will be developed into two papers for publication.

2. b. iv CONTROL OF HAND AND WRIST

Abstract

The primary progress in this quarter was to continue work on the software described in the last QPR. The basic structure has been maintained, but we have added flexibility with regard to the order in which different components can be executed, and the way in which data is stored and retrieved. We attempted to complete one experiment, but discovered a problem with how the stimulus frequency was controlled that prevented any useful data collection. This problem has been corrected.

Purpose

The goal of this project is to design control systems to restore independent voluntary control of wrist position and grasp force in C5 and weak C6 tetraplegic individuals. The proposed method of wrist command control is a model of how control might be achieved at other joints in the upper extremity as well. A weak but voluntarily controlled muscle (a wrist extensor in this case) will provide a command signal to control a stimulated paralyzed synergist, thus effectively amplifying the joint torque generated by the voluntarily controlled muscle. We will design control systems to compensate for interactions between wrist and hand control. These are important control issues for restoring proximal function, where there are interactions between stimulated and voluntarily controlled muscles, and multiple joints must be controlled with multijoint muscles.

Progress Report

Details of continuing software development will not be presented. The last quarterly progress report presented the overall structure. Results of software testing and demonstration will be presented in the next quarterly report.

Plans for next quarter

We plan a new set of experiments at the beginning of the next quarter. We expect that we will need to add further enhancements to the software, particularly visualization of the data related to performance during the experiments.